

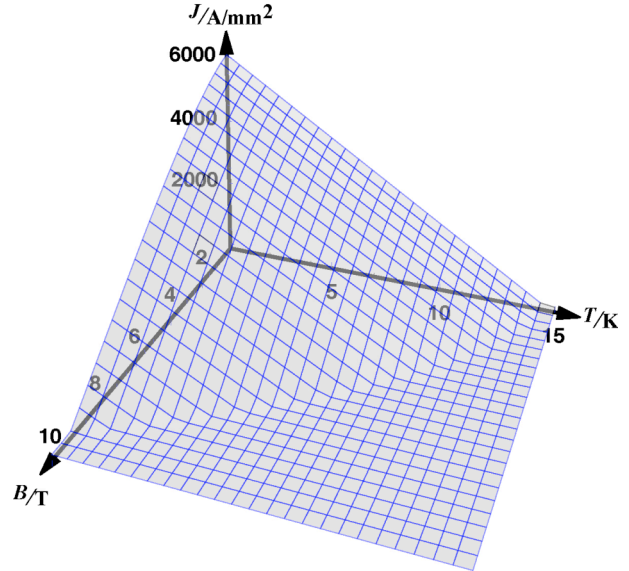
# Impediments to High Field MR – a Look at $B_0$ and $B_1$ Field Behaviour

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## $B_0$ Field Behaviour

The most obvious impediment to performing MR on humans at ever higher field strengths is the cost, size and complexity of superconducting magnets. The primary culprit here is the inability of superconducting wire to carry large currents when in a large magnetic field, as shown in Fig. 1. Thus, as field strengths increase, magnets must have more wire to create the desired field, and so become larger and more expensive. Further, because high currents are now also no longer permitted at the *ends* of the superconducting solenoid where turns tend to be concentrated to obtain good homogeneity, the device must become longer to maintain that homogeneity. In turn, the forces involved between conductors become increasingly large, necessitating greater constructional ruggedness. To progress to even higher fields (greater than say 8 T), one must either further cool the wire (less than the usual 4.2 K, the temperature of liquid helium) and/or switch to brittle alloy superconductors – even more expensive propositions. (Note that so-called “high-temperature” superconductors, functioning in liquid nitrogen at 77 K, cannot currently support MRI field strengths.)



**Figure 1.** A typical “critical surface” for superconducting wire. The surface is a function of field strength  $B$  in tesla, temperature  $T$  in kelvin and current density  $J$  in Ampère/mm<sup>2</sup>. Above the surface, the wire is “normal”, i.e. it has resistance. Only behind the surface is it superconducting.

## SAR and Chemical Shift

From a physicist’s point of view, at a more basic level the behaviour of the  $B_0$  field in humans is no different at high field from that at low field. The various organs in the human body have their diverse magnetic susceptibilities that distort the field in a manner that is *proportional* to field strength. Thus, the thyroid gland, for example, produces  $B_0$  field inhomogeneities in its vicinity that are exactly the same in *parts per million* (ppm) at a field of 0.1 T as they are at 10 T. Likewise, chemical shifts remain constant in ppm and the frequency spread of a spectrum *in these units* does not change with field strength.

Problems arise from an MR point of view, however, because the various parameters we consider important ( $1/T_1$ ,  $1/T_2$ , specific absorption rate [SAR], signal-to-noise ratio [S/N] etc.) are *not* fractionally constant with change of field strength – we must work in absolute rather than fractional units. Thus, while the relative inhomogeneity of the  $B_0$  field across the thyroid may well be constant at, say, 2 ppm, the *absolute* spread in Larmor frequency is  $2\nu$  Hz, where  $\nu$  is the Larmor frequency in MHz. In other words, the spread is proportional to  $B_0$ . Likewise the chemical shift  $\sigma$  ppm is  $\sigma\nu$  in Hz, and to cover these bandwidths, a  $B_1$  field that increases linearly with frequency, i.e. field strength  $B_0$ , is needed. This in turn implies that the required RF transmitter power increases as roughly  $B_0^2$ . However, things are actually much worse than this! When we look at where the transmitter power goes, we find that it is absorbed in the human body as heat, and to a first approximation the efficacy of this heating mechanism is also proportional to the square of the Larmor frequency. The power specification for the RF transmitter then increases frighteningly quickly as roughly  $B_0^4$ , and before long, we run into SAR problems, for allowing for the decrease of pulse length with increasing  $B_1$  field strength, SAR increases as  $B_0^3$ , all other factors such as pulse repetition rate, number of slices, etc. remaining constant. To put these dependencies in perspective, if 1 W of power is needed at 0.1 T to produce a 90° pulse, 100 megawatts would be needed at 10 T to cover the same chemical shift range! This is clearly ridiculous, and so the first casualties here are generally the number of slices collectable followed by our

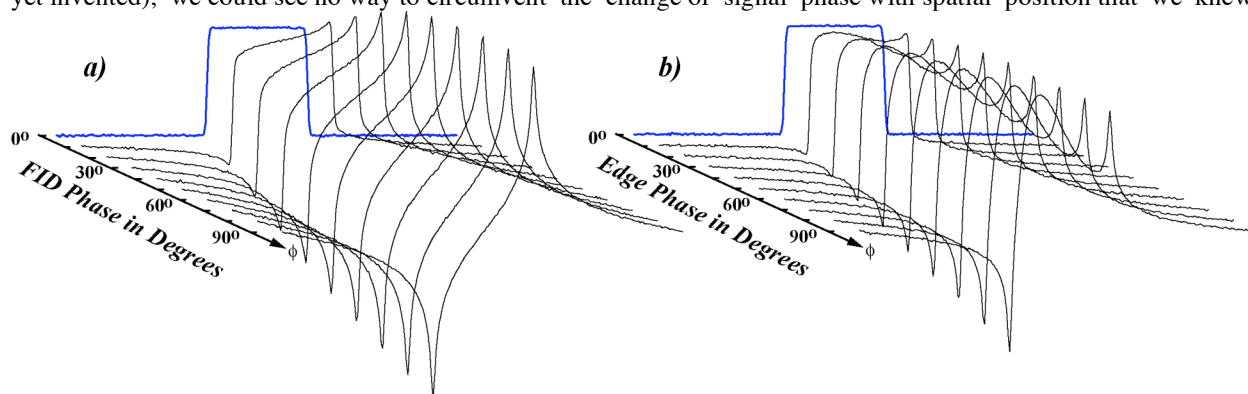
ability to cover the entire chemical shift range. We are forced to excite only water, or some other limited spectral range.

### *S/N and Chemical Shift Artefacts*

Looking at the time-domain signal, the duration  $T_2^*$  of, for example, the thyroid FID (or echo) is of the order  $1/(2\nu)$  seconds – the echo becomes shorter and shorter with increasing field, while the true  $T_2$  alters but little. Much of the FID or echo signal is therefore lost because of dephasing. If one does not shorten the data acquisition window to accommodate this phenomenon, the result is distortion in the image round the thyroid, or indeed round any other magnetically susceptible piece of anatomy. A related problem concerns chemical shift again. If we assume a chemical shift of 3.5 ppm between water and fat, the acquisition window must shrink in time as  $1/(3.5\nu)$  (i.e. inversely with  $B_0$ ) if water and fat in the same physical space (voxel) are not to appear in the image as being displaced relative to one another – the “chemical shift artefact”. This implies that at 64 MHz, for example, the acquisition window should be no more than about 4 ms, while at 128 MHz it should be only about 2 ms. The immediate consequence is that the S/N of an image does not increase as rapidly with increasing  $B_0$  as one might expect – typically only as the square root of field strength. The only way to get the lost signal (and S/N) back is by forming multiple spin echoes. Alternatively, and as a compensation, we may be able to squeeze more slices into the time that has been released, but with too many echoes *or* slices, we again we run into SAR problems – we simply cannot safely apply too many pulses. Of course, if there is *no* fat in the anatomy of interest, one can open up the acquisition window until distortion just becomes noticeable, and the S/N and/or resolution in the image can then look marvellous. However, the same imaging parameters with an abnormal subject who *has* fat will result in an image with clear chemical shift artefacts. Beware – this is fertile ground for salesmanship!

### **$B_1$ Field Behaviour**

Now none of this is new. In the early days of imaging, the “optimal field strength” was a “hot” topic for debate and the author was asked to give an address to the 1984 SMRM conference on the controversy. The resulting papers (1) examine in detail this whole issue of the effects of  $B_0$ , of restrictions on SAR and hence  $B_1$ , etc., and they are still directly relevant: they cover much more than is possible here. However, what change above roughly 80 MHz are the premises underlying some of the calculations made in those and other papers. Hoult and Lauterbur (2) (H & L), in their classic paper on MR S/N, assumed that a quasi-static solution of Maxwell’s equations sufficed. In other words, we assumed that at the frequencies used in human MR, the wavelength  $\lambda$  of radio waves in the human body was much bigger than the size of the body and that radio waves therefore played a negligible role in MRI. (For a paper discussing the origins of the myth of radio waves as an explanation for MRI signal reception at common field strengths and sample sizes, see (3).) There were two reasons for these assumptions: to be honest, I didn’t have the mathematical expertise as a young post-doc. to solve Maxwell’s equations for a human model, and secondly, as imaging then relied on Fourier-transformation of a correctly-phased *FID* (transforming a symmetrical echo was not yet invented), we could see no way to circumvent the change of signal phase with spatial position that we knew



**Figure 2.** The effects of incorrect phase during Fourier transformation of the FID from a slice through a cubic water phantom with the read gradient on. The rear “top-hat” waveforms (heavier [blue, if coloured] lines) are the correct 1-D projections. The distortions to the projection become progressively greater as the phase error increases and a dispersive signal begins to replace the absorptive one. In *a*), the phase alters for the entire projection – an instrumental or reconstructive mis-setting. In *b*), the phase changes as the square of distance from the middle of the cuboid, an unavoidable high frequency effect, see Fig. 3. Use of a symmetrical echo, rather than an FID prevents the generation of a dispersion curve during Fourier transformation, and so removes the artefact. However, a magnitude image must be taken to retain correct image intensities.

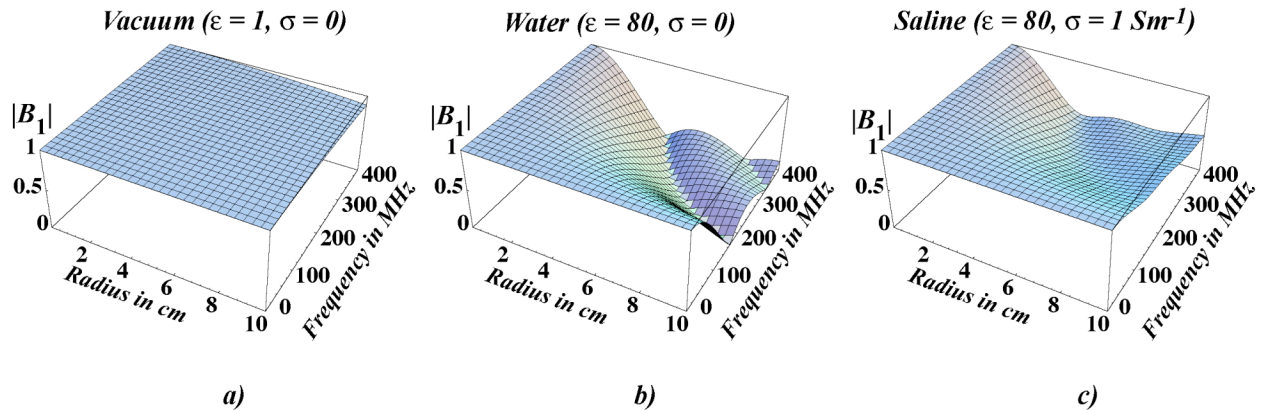
would occur at “high frequency”. With such phase changes, the direct relationship between proton density/spatial position and Fourier-transformed signal strength/frequency breaks down, as shown in Fig. 2. Thus we felt that imaging above 10 MHz was not likely to be viable! In fact, thanks to the great idea of Fourier transformation of an echo rather than the FID (4), the H & L theory has been useful up to roughly 70 MHz because it predicts signal strength with no reference to phase; the former tends to change more slowly with frequency than the latter. However, the first cracks in it showed up in the form of asymmetric image brightness as long ago as 1985 (5). The asymmetry, which has been noted many times since, is a fact of life in many high-field MRI experiments, but causes much head-scratching (see below). Far more serious cracks appeared with the head images obtained at 8 T by Robitaille *et al* (6). There is a (now well-known) bright area in the image centre that H & L cannot explain, and Robitaille proposed in several talks his own, rather exotic, theories.

### Penetration Effects

Transformed echoes are now part of MRI history, I have improved my mathematics, and a full theory of S/N is available (7) that extends H & L to ultra-high frequencies. Sadly, but not surprisingly, the pristine simplicity of the H & L theory has been lost. However, the conclusions of the paper are readily accessible (the article is designed so that the mathematics can be skipped) and they hold surprises. While the complexity of the human body means that computer simulation (8) is the only way accurately to model the  $B_1$  field spatial variation, as always an analytical model yields great insight. As in H & L, I therefore used as a substitute for the human head a phantom comprising a mildly conducting saline sphere whose conductivity and permittivity could be varied. The first point that appeared (and this has met considerable resistance) is that, try as one might, a homogeneous  $B_1$  field can *not* be generated: it is not a solution of the fundamental tenets of electromagnetism: Maxwell’s equations. It is actually very simple to prove this in one line of mathematics. A radio-frequency magnetic field  $\mathbf{B}$  oscillating or rotating at angular frequency  $\omega_0$  *must* obey the Helmholtz equation (7), which is derived from Maxwell’s equations:

$$\nabla^2 \mathbf{B} + k^2 \mathbf{B} = 0; k^2 = \mu\mu_0\epsilon\epsilon_0\omega_0^2 - i\omega_0\mu\mu_0\sigma \equiv \left(\frac{2\pi}{\lambda}\right)^2 - i\omega_0\mu\mu_0\sigma$$

Here  $\mu$  ( $\sim 1$ ),  $\epsilon$  ( $\sim 50$ ) and  $\sigma$  ( $\sim 0.5$ ) are respectively the magnetic permeability, electric permittivity and conductivity of the human body while  $\mu_0$  and  $\epsilon_0$  are the permeability and permittivity of free space. The wavelength in the human body, as earlier, is  $\lambda$ . If magnetic field  $\mathbf{B}$  is homogenous, then it has no variation with spatial distance and so  $\nabla^2 \mathbf{B} = 0$ . The Helmholtz equation is then only obeyed when  $k^2 = 0$ ; i.e. when the Larmor angular frequency  $\omega_0$  is zero. In other words, a homogeneous field is only a static solution, and the more the wavelength  $\lambda$  shrinks and approaches the dimensions of the human body as  $B_0$  field strength rises, the more inhomogeneous the RF field *must* become, as shown in Fig. 3. (This explains the bright central area that one sees in high-field head images. The field is “focussed” in the middle of the head, and contrary to some explanations, this has nothing to do with a resonance phenomenon.) It is possible to obtain a homogeneous field over certain surfaces (e.g. a transverse plane), but novel



**Figure 3.** The theoretical variation (to zeroth order only) of the  $B_1$  field strength with radius and frequency in a spherical sample when the surrounding RF coil has been designed to give a homogeneous field (7). With no sample in the coil (a), the field strength hardly varies with radius, even at 400 MHz. However, in water (b) with its high dielectric constant ( $\sim 80$ ), there is substantial radial dependence of  $B_1$  even at 100 MHz, giving a darker, modulated image away from the centre. This effect is mitigated, but not eliminated, by the addition of salt (c) – a reasonable model for the human head’s electrical properties.

strategies, such as a sequence of excitation pulses (with high SAR?), will be needed to generate a uniform flip angle over an arbitrary volume. Even then, the phase of excitation is unlikely to be spatially invariant.

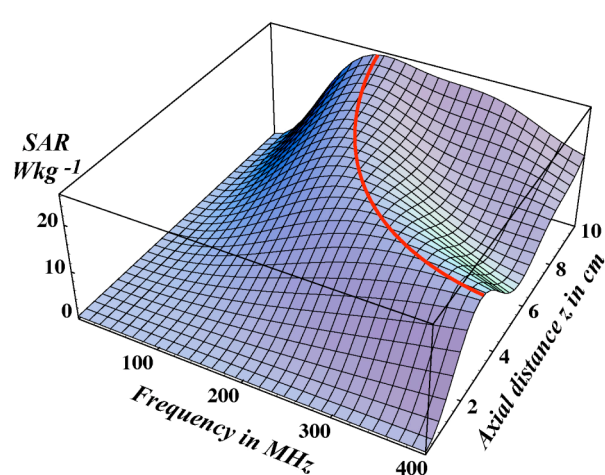
### RF Coil Problems

Another problem is that conventional excitation coils become inefficient and generate of themselves an inhomogeneous  $B_1$  field *in vacuo* before a sample is even considered. As soon as their conductor length approaches a fraction  $\alpha \sim 1/10$  of a wavelength *on the wire surface*, the propagation of a wave of electrical current down the wire takes a significant time. Thus, as there is generally some sort of reflection at the ends of the wire where the electrical environment changes, the *equilibrium* distribution of current amplitude along the wire may be uneven – for example, as  $\cos(2\pi x/\lambda)$ ,  $-\alpha\lambda/2 < x < \alpha\lambda/2$ , where  $x$  is position on the wire. It follows that the field of Fig. 3a, though theoretically feasible, is not easy to generate. Conductors have to be broken up into an array of smaller coils (a “phased array”) to allow effective generation of a desired field – e.g. a homogeneous field in a transverse head slice. Worse, interaction between the coils, via the intermediary of the sample, spoils our ability to generate what few homogeneous fields *are* feasible. Current in one coil generates a  $B_1$  field. This causes RF current to flow in the sample which in turn generates another RF field which then induces voltages in the other coils. These voltages alter the coil currents from those desired, which changes their  $B_1$  fields and so on, in a nasty interactive brew. Fortunately, a solution for this interaction problem appears at least to be in sight (9): it has been shown that a coil’s RF current may be compared in amplitude and phase “on the fly” with what was intended, and corrected by the mechanism of negative feedback.

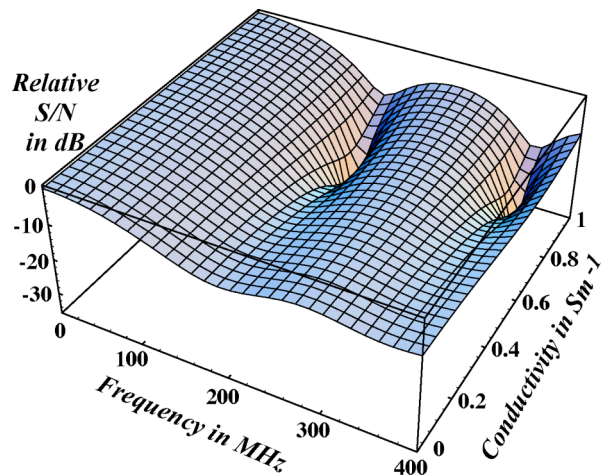
When *receiving* signals with such an array, the individual component signals can be massaged in a computer to give an overall image that appears homogeneous in brightness. It is perhaps this that has led people to assume that upon transmission a homogeneous  $B_1$  field can be obtained. As already stated, this cannot be, for the fields from the individual coils add vectorially and there is nothing we can do about it – the sum field must obey the Helmholtz equation.

### SAR

Associated with the sample-mediated inhomogeneous  $B_1$  field shown in Fig. 3c is a change in the distribution of the RF *electric* field  $\mathbf{E}$ . (The electric field is directly associated with power deposition:  $\text{SAR} \propto \mathbf{E}^2$ .) As a result, with increasing frequency the region of maximum SAR tends to move inside the sample, as shown in Fig. 4. Extrapolating the phantom results to a human, as perfusion may not be as robust inside the head as directly under the skin, we may need to re-examine our SAR assumptions. A benefit of the change in the  $\mathbf{E}$ -field distribution is that the *total* power needed for a pulse does not increase quite as rapidly with  $B_0$  as one might be led to believe from H & L. Concomitantly, the average S/N increases somewhat more rapidly than one might expect from H & L.



**Figure 4.** A plot of specific absorption rate versus frequency and axial distance  $z$  for a sphere of saline ( $\sigma = 1\text{S/m}$ ) subject to a rotating transverse  $B_1$  field of  $5.87\text{ }\mu\text{T}$  ( $\nu_1 = 250\text{ Hz}$ ). The region of maximum SAR moves into the sample (thicker [red if coloured] line) and decreases slightly as the frequency increases.



**Figure 5.** The signal-to-noise ratio of the FID from the *entire spherical sample* as a function of frequency and conductivity, relative to that predicted by H & L. The sample radius is 10 cm. There are two deep “holes” in the plot caused by destructive interference associated with diverse voxel signal phases over the sample.

### Diverse Details

Finally, some arcane points that come out of the analysis: 1. If the  $B_1$  field during transmission is calculated in the positively rotating frame, the received signal is then found from the complex conjugate of the hypothetical, unit-current, receptive  $B_1$  field in the *negatively* rotating frame (10). This subtlety is particularly important in computer simulations. For example, it is difficult to see how a transmitting/receiving system that has symmetry (e.g. a circular surface coil whose axis is in the  $x$  direction) could generate sagittal images with asymmetric brightness (5). However, an analysis with the correct rotating frames confirms this phenomenon. The symmetry is broken by the handedness of magnetic resonance precession. 2. With certain combinations of sample conductivity, permittivity and size, it is theoretically possible with a phantom to receive zero signal after a  $90^\circ$  pulse as shown in Fig. 5, thanks to a spread of phases across the sample! To the best of the author's knowledge, this phenomenon has not yet been observed, but it is wise to keep alert to the possibility of such odd behaviour.

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